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A Stationary-Sources and Rotating-Detectors Computed Tomography Architecture for Higher Temporal Resolution and Lower Radiation Dose

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ABSTRACT In current computed tomography (CT) architecture, both X-ray tubes and X-ray detectors are rotated mechanically around an object to collect a sufficient number of projections. This architecture has been shown to not be fast enough for patients with high or irregular heart rates. Furthermore, both X-ray beams and detectors of the current architecture are made wide enough, so that the entire object is covered in the lateral direction without data truncation. Although novel acquisition protocols have recently been developed to reduce a radiation exposure, the high radiation dose from CT imaging remains a heightened public concern (especially for cardiac CT). The current CT architecture is a major bottleneck to further increase the temporal resolution and reduce the radiation dose. To overcome these problems, we present an innovative stationary-sources rotating-detectors CT (SSRD-CT) architecture based on the three stationary distributed X-ray sources and three smaller rotating X-ray detectors. Each distributed X-ray source has $\sim$100 distinctive X-ray focal spots, and each detector has a narrower width compared with the conventional CT detectors. The SSRD-CT will have a field-of-view of 200 mm in diameter at isocenter, which is large enough to image many internal organs, including hearts. X-rays from the distributed sources are activated electronically to simulate the mechanical spinning of conventional single-beam X-ray sources with a high speed. The activation of individual X-ray beam will be synchronized to the corresponding rotating detector at the opposite end. Three source-detector chains can work in parallel to acquire three projections simultaneously and improve temporal resolution. Lower full-body radiation dose is expected for the proposed SSRD-CT because X-rays are restricted to irradiate a local smaller region. Taken together, the proposed SSRD-CT architecture will enable $\leq$50-ms temporal resolution and reduce radiation dose significantly.

INDEX TERMS Computed tomography, temporal resolution, radiation dose, multisource, interior tomography.

I. INTRODUCTION

Since its Nobel-prize winning invention in 1970s [1], x-ray computed tomography (CT) has been through dramatic developments, and the performance of CT scanners has been improved tremendously. One major improvement worthy of mention is temporal resolution, which has been improved from 5 min at the beginning to sub-second at this moment [2]. The current highest temporal resolution is achieved by a second-generation dual-source CT (DSCT, e.g. Definition Flash CT from Siemens), providing a temporal resolution up to 75ms [3], [4].

The need for a higher CT temporal resolution stems from an obvious fact that organs within human bodies are constantly moving. In particular, because heart is the fastest moving organ within the human body, CT developments have been largely directed at improving temporal resolution for cardiac CT. A cardiac CT scan with higher temporal resolution can 'freeze' heart motion more effectively, thus it provides sharper images and fewer artifacts. Although it is favorably reviewed in many clinical cardiac CT imaging, the state-of-the-art temporal resolution of 75ms works best only when image acquisition is ECG-gated to the relatively quiet diastolic phase [5]–[9]. To image various cardiac phases from patients with irregular heart rates (e.g. arrhythmia), it was estimated that a temporal resolution down to 50ms would be needed [10], [11].

To achieve a higher temporal resolution, researchers have previously investigated two strategies. The first and more obvious strategy is to reach a faster gantry rotation speed. This has been the primary strategy in action during the last few decades. Many folds of improvement in gantry rotation speed has been achieved since 1980s. Currently, the fastest gantry rotation speed is about 300ms. However, increasing the temporal resolution via an even faster gantry rotation speed appears to be beyond today's mechanical limits. Nowadays, the gravitational force on the gantry can reach 30g, about 10 times higher than the acceleration experienced during the space shuttle take-off [2], [12]. An innovative idea for ultrafast CT is the electron beam CT (EBCT), which employs a stationary target ring and a stationary detector ring. Because EBCT sweeps an electron beam across a stationary anode ring electronically (i.e. without mechanical gantry rotation) [13], it can achieve a temporal resolution of about 30ms.

The second strategy is to acquire multiple projections simultaneously through multiple source-detector imaging chains. This strategy is well known to improve the temporal resolution of CT scanners. For example, the 75ms temporal resolution of the DSCT is achieved by arranging two x-ray tubes and two detectors on a single gantry with a gantry rotation time of 270ms. Previous simulations showed that doubling the number of imaging chains on a given gantry is more efficient to increase the temporal resolution than reducing the gantry rotation time by a factor of 2 [14]. However, as demonstrated by the dynamic spatial reconstructor (DSR) in the 1980s and depending on the actual geometry, the number of imaging chains was limited by the proximity of adjacent imaging chains to prevent adjacent x-ray beams from overlapping [15]–[17].

Another important improvement in CT is reduction in radiation dose. The radiation risk (e.g. cancer) from CT imaging is a growing public concern [18]. In clinical practices, it generally requires adhering to the principle of reducing radiation doses as low as reasonably achievable (ALARA). The dose concern is particularly critical for CT perfusion imaging, where a high radiation dose ($\sim$10mSv) can be accumulated from multiple CT acquisitions [19]. The dose issue will become even more prominent if a higher spatial resolution is required. This is because, to double the spatial resolution in all 3 spatial directions without affecting the contrast-to-noise ratio (CNR), the radiation dose has to be increased by a factor of 2^4 (i.e. 16-fold increase) [20]. Although a number of dose reduction techniques have been developed [21]–[25], more works are needed to allay public concern on CT radiation dose.

Aiming to improve temporal resolution and lower radiation dose in this paper we propose a new CT architecture by combining the latest compressive-sensing (CS) based interior tomography [26] and distributed x-ray source technology [27], [28]. The interior tomography technique allows accurate reconstruction of a region-of-interest (ROI) from truncated projections. In contrast to the conventional CT architecture, the detectors for interior tomography do not cover the full transaxial extent of the object. The interior tomography technique can help to reduce detector size, suppress scattering artifacts, and lower radiation dose. Furthermore, radiation doses could be further reduced when interior tomography is combined with the compressive-sensing framework [26]. A distributed x-ray source can produce x-rays by extracting electron beams from an array of cathodes and sending each electron beam to a distinctive focal spot on the same target. The cathodes are made of either hot dispenser cathode emitters (DCE's) [27] or cold field emitters such as carbon nanotubes (CNT's) [28]. Electron beams (and hence the corresponding x-ray beams) can be switched on and off electronically by applying the corresponding extraction voltages on the cathodes. By programming these extraction voltages, a scanning x-ray beam can be generated to irradiate an object from different angles, enabling tomographic imaging without mechanical motion [29]. The CNT-based distributed x-ray sources are very attractive because the cold CNT cathodes work at room temperatures and allow compact packaging and faster and more flexible control of x-ray beams [30].

In the rest of this paper, we will first present our new architecture, and then demonstrate its feasibility with numerical analyses and simulation using a clinical cardiac CT dataset. The results will be presented and the rationales for the expected performance will be discussed. Finally, we will discuss the potential limitations and challenges for the implementation of the proposed SSRD-CT architecture.

II. THE PROPOSED CT ARCHITECTURE

The proposed CT architecture is illustrated in Fig. 1. It has three stationary distributed x-ray sources and three rotating detectors. Hereafter, we refer this architecture as stationary-source rotating-detector CT (SSRDCT). Three identical source-detector chains are positioned around an object symmetrically. The key geometrical parameters are listed in Table 1. The source-to-isocenter distance (SID) and the source-to-detector distance (SDD) are 540mm and 950mm, respectively, which are close to most of commercially available CT scanners. The field-of-view (FOV) is at 200mm, which is sufficient to cover the transaxial extent of a typical human heart. This design will allow us to reuse, or use with slight modifications, many existing and mature CT technologies in detectors, gantry, slip rings, etc. The core

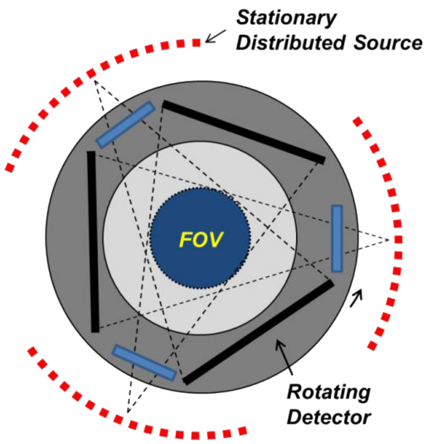

FIGURE 1. Schematics of the proposed CT architecture with three stationary distributed x-ray sources and three rotating detectors.

TABLE 1. Geometrical parameters of the proposed CT architecture.

Number of Source-Detector Chains	3
Source to Isocenter Distance (mm)	540
Source to Detector Distance (mm)	950
Field of View	200

new elements are the interior tomography and the distributed x-ray sources.

Each distributed x-ray tube will have a multitude of x-ray sources that are distributed equiangularly (relative to the isocenter) in a shared vacuum envelope. During a CT scan, the x-ray sources within an x-ray tube will be sequentially turned on and off from one end to the other. As a result, each x-ray source will give a pulsed radiation, and the switching is done through programming the gate voltage on the corresponding cathodes [27], [30]. Furthermore, the switching of x-ray sources will be synchronized with to the rotation of the detectors to simulate the spinning of three traditional x-ray sources.

All three x-ray beams will be collimated toward the 200mm FOV at isocenter by three collimators that are mounted on the gantry between the sources and detectors. The collimated beams can reduce radiation exposure to regions outside the FOV. The 200mm FOV is much smaller compared to the 500mm FOV in most commercially available CT scanners, but it should be enough to cover many important organs such as hearts within human bodies. Smaller FOV also allows a smaller detector size (~60% reduction). Smaller detector sizes make it possible to mount three imaging chains on a conventional CT gantry (~1m in diameter) without overlapping. As a result, three projections can be acquired from the three imaging chains simultaneously.

III. NUMERICAL ANALYSES

A. SCAN ANGLE

The three identical source arrays are symmetrically positioned around the FOV (see Fig. 2). To acquire a complete

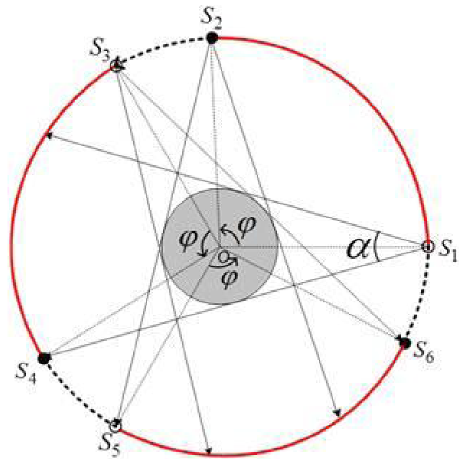

FIGURE 2. Illustration for scan angle analysis of the three distributed x-ray source (red color). The three source arrays are arranged symmetrically around the FOV, which is shown as the inner circular region (grey color).

projection dataset for the FOV, the effective scan angle should be greater than $\pi + \alpha$ where α is the fan angle and indicated by the angle $\angle S_1OS_4$. Suppose each source array can cover an angle φ,

$$\angle S_1OS_2 = \angle S_3OS_4 = \angle S_5OS_6 = \varphi. \tag{1}$$

Then we have

$$\angle S_2OS_3 = \angle S_4OS_5 = \angle S_6OS_1 = (2\pi - 3\varphi)/3. \tag{2}$$

For a short scan (i.e. $\pi + \alpha$) from S_1 to S_4, the data from S_2 to S_3 are missing. However, the missing data can be compensated by the source array from S_5 to S_6. Therefore, we have

$$\begin{cases} \angle S_1OS_4 &= \pi + \alpha \\ \angle S_1OS_4 &= \angle S_1OS_2 + \angle S_2OS_3 + \angle S_3OS_4 \\ &= \varphi + (2\pi - 3\varphi)/3 + \varphi. \end{cases} \tag{3}$$

We immediately arrive at

$$\varphi = \pi/3 + \alpha. \tag{4}$$

As a result, to satisfy the data requirement for a short scan, each source array should cover an angle of $\pi/3+\alpha$, and the corresponding detector need to be rotated an angle of $\pi/3+\alpha$. From the key geometry parameters listed in Table 1, the fan angle α can be determined to be 21°.

B. TEMPORAL RESOLUTION

Based on the above analysis in subsection III.1, the temporal resolution can be easily determined for a short scan in the SSRDCT architecture as $(\pi/3+\alpha)/2\pi * T_{rot}$, where T_{rot} is the gantry rotation time. In contrast, the temporal resolutions for short scan of the DSCT and the conventional single-source CT (SSCT) architectures are $(\pi/2+\alpha)/2\pi * T_{rot}$, and $(\pi + \alpha)/2\pi * T_{rot}$ respectively. For half scan (i.e. 180° scan), the temporal resolution would be $T_{rot}/6$, $T_{rot}/4$, and $T_{rot}/2$ for SSRDCT, DSCT, and SSCT, respectively. The relative

TABLE 2. Comparison of relative temporal resolutions for the SSCT, DSCT, and STRICT architectures. The gantry rotation time was assumed to be same and the SID is 540 mm.

Architecture	SSCT	DSCT	SSRDCT
Number of Source-Detector Chains	1	2	3
Field of View (mm)	500	340[a]	200
Relative Temporal Resolution – Short Scan[b]	1	0.54	0.35
Relative Temporal Resolution – Half Scan[b]	1	0.50	0.33

[a] The two detectors in DSCT have different sizes and the system FOV is limited by the smaller detector at 340mm [4].

[b] Temporal resolutions are normalized to those in the traditional SSCT architecture.

temporal resolutions for the three architectures are listed in Table 2, where the gantry rotation time T_{rot} was assumed to be same and the SID was taken as 540mm. Compared to the latest DSCT architecture, the temporal resolution of SSRDCT architecture is improved by a factor of more than 1/3 for both the short-scan and half-scan modes. If the gantry rotation time T_{rot} is assumed as 270ms, the temporal resolution of the SSRDCT architecture will be 61ms for the short-scan mode and 45ms for the half-scan mode.

C. RADIATION DOSE

At the same SID, the radiation dose at isocenter is proportional to the total radiation exposure during a CT scan. For a same SID, a constant x-ray flux can be assumed at the isocenter for a given x-ray source technology. Therefore, the radiation dose at isocenter is solely dependent on the total exposure time. For the SSCT architecture, the x-ray source is turned on continuously during the data acquisition for a single slice image, thus the total exposure time per CT scan equals to the corresponding temporal resolution. For the DSCT architecture, since the two sources are turned on continuously during the data acquisition, the total exposure time is two times of its temporal resolution. For the SSRDCT architecture, each x-ray source will give a single pulsed exposure, thus the total exposure time is the summation of x-ray exposure times from individual x-ray sources in the three source arrays. This total exposure time is not necessarily three times of its temporal resolution, because an 'off time' (i.e. a time period without exposure) can be introduced between two consecutive pulsed exposures from two neighboring x-ray sources. The off time will significantly reduce the total exposure time per CT scan in the SSRDCT architecture. This can help save the radiation dose.

To reduce radiation exposure to patients, many efforts have been devoted to develop image reconstruction algorithms for fewer projection views, less of a scanning arc or lower x-ray intensity per view [26], [31], [32]. These algorithms are highly synergistic to our proposed architecture. They would permit a decent image quality to be obtained in the SSRDCT architecture with less x-ray sources in each source array, smaller coverage angle per array, and lower x-ray source power, resulting in a CT system that is more compact and at lower cost. Particularly, lowering x-ray intensity per view will enable the stationary distributed x-ray sources to adopt the fixed-anode design, which will dramatically simplify the source design and reduce the demand on the amount of electron currents from the cathodes. In the following section, we will evaluate the performance of the architecture with a clinically acquired cardiac CT dataset and the popular total variation minimization (TVM) method combined with steepest descent (TVM-SD) search [33], [34].

IV. SIMULATION AND RESULTS

To evaluate the image quality of proposed CT architecture, we carried out numerical simulations with clinically acquired cardiac CT imaging data in a typical fan-beam geometry. The data were acquired on a commercial CT scanner (GE Discovery CT750 HD) at Wake Forest University Health Science. The radius of the scanning trajectory is 538.5mm. 2200 projections were uniformly acquired over a full scan of 360°. Each projection has 888 equiangular distributed detector elements. The popular total variation minimization (TVM) with steepest descent (TVM-SD) search algorithm [33], [34] was used to reconstruct a global image (Figure 3 (a)), which serves as a benchmark to evaluate the image quality of the SSRDCT architecture. The global

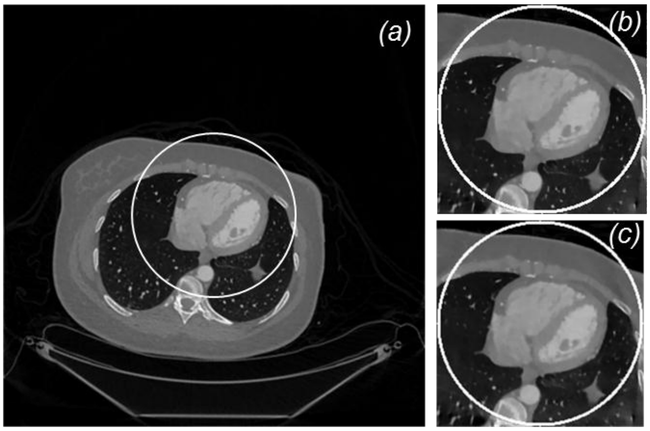

FIGURE 3. Reconstructed slice images using the TVM-SD algorithm with a display window [−1000HU, 1800HU]. (a) Is the reconstructed global image from full dataset after 60 iterations, (b) and (c) are the reconstructed ROI images for the short-scan mode and half-scan mode in the SSRD-CT architecture after 200 iterations.

image is a matrix of 512 × 512 pixels and covers a region of 498.3 × 498.3mm^2.

To simulate an interior-ROI-oriented data acquisition, each projection was truncated by discarding 270 detector elements from each of the two sides. This resulted in an interior ROI of 200mm in diameter with a fan angle of 21.4°. Therefore, for a short scan the cover angle of each source array should be 81.4°, and only the acquired projections in [0, 81.4°], [120°, 201.4°], and [240°, 321.4°] are needed in the SSRDCT architecture. Correspondingly, for a half scan only the projections in [0°, 60°], [120°, 180°], and [240°, 300°] are needed. These selected projections were first truncated, and then used to reconstruct interior images using the TVM-SD algorithm. Figure 3(b) and 3(c) show the reconstructed ROI images for the short-scan and half-scan modes, respectively. Figure 4 shows the representative line profiles along the horizontal and vertical central lines inside the ROI in the three images in Figure 3. We can see there is an excellent match between the global image and the ROI images.

The TVM-SD algorithm was developed based on the CS theory. According to the CS theory, high-quality images can be reconstructed from a limited number of projections. This means fewer views can be used (thus reducing radiation dose). For the half-scan mode (i.e. 60° coverage angle per source array), we down-sampled the projections using the global benchmark image. The projections views for the x-ray sources in the three source arrays were equiangularly spaced in the [0, 60°], [120°, 180°] and [240°, 300°] angular ranges. The number of views per source array was set to be 160, 130, 100, 70 and 40, respectively, and the corresponding reconstructed ROI images are shown in Figure 5. Figure 6 shows two representative line profiles along horizontal and vertical central lines. From Figures 5 and 6, we can see that the reconstructed images from 160, 130 and 100 views per array are all qualitatively comparable with the benchmark, while the others lose some details as indicated by the arrows in Figure 5 (d) and (e).

To quantitatively evaluate the quality of reconstructed images from the half-scan mode, we calculated the root-mean-square error (RMSE) in the ROI region between the reconstructed ROI images and the benchmark. The results are shown in Figure 7. We can see that when the iteration number reaches 1000, the RMSEs of 160 × 3, 130 × 3 and 100 × 3 views are significantly smaller than those of 70 × 3 and 40 × 3 views. Its implication for dose reduction is highly significant. If we use the result from 100 × 3 views as a reference, this means that the radiation dose could be potentially reduced by a factor of 3.7.

V. DISCUSSION AND CONCLUSION

In this paper, a SSRDCT architecture is presented by combining two latest advances in the CT field: the interior tomography [26] and the distributed x-ray source [27], [28]. The feasibility is investigated via extensive numerical analyses and simulations using clinical cardiac CT data. Compared to the current 3rd-generation CT architecture, the SSRDCT has a great potential to reach ≤50ms temporal resolution

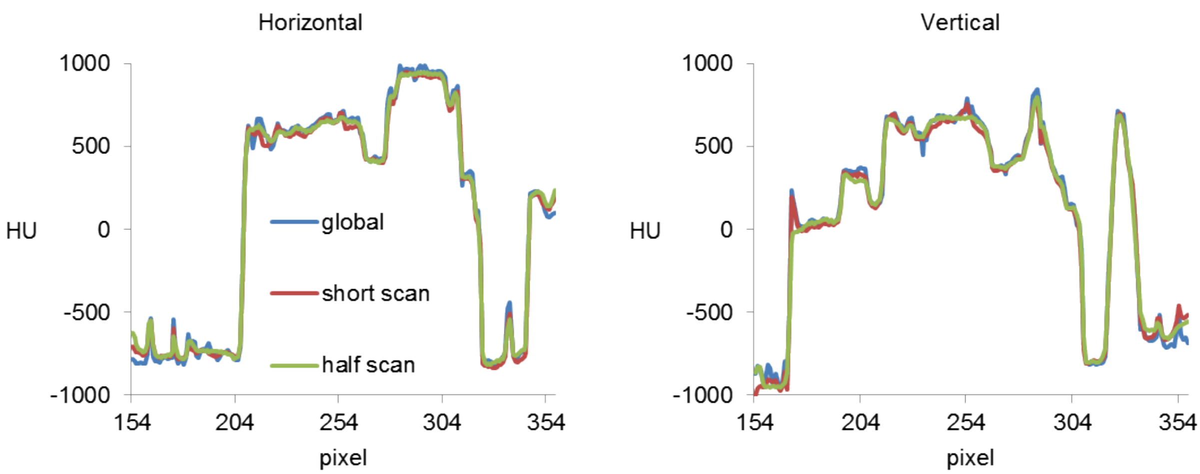

FIGURE 4. Representative line profiles of the images in Figure 3 along the horizontal and vertical directions at the center.

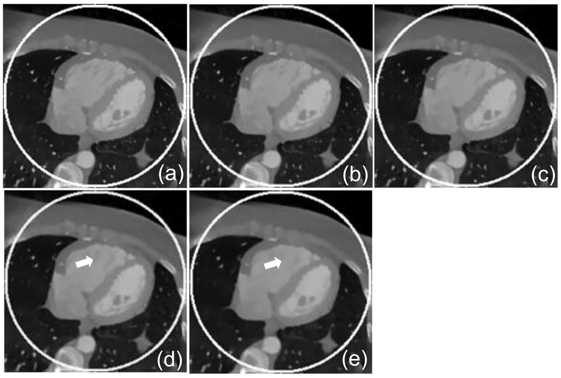

FIGURE 5. Reconstructed ROI images for the half-scan mode with few views in a display window [−1000HU, 1800HU]. The Iteration numbers for all the images are 1000, and the total number of views are 160 × 3, 130 × 3, 100 × 3, 70 × 3, and 40 × 3 for (a)–(e), respectively.

and reduce radiation dose significantly. The enhanced performance metrics have high diagnosis values for clinical exams especially for cardiovascular diseases (CVDs). Indeed, cardiac CT has been used to help diagnose a range of CVDs including coronary heart disease, valvular heart disease, pericardial diseases, congenital heart disease, and pulmonary vein disease [35].

Previously, high CT temporal resolution has been sought out by two approaches - DSR and EBCT. While fast temporal resolution in the DSR was achieved by increasing the number of source-detector chains, for the EBCT, fast temporal resolution was obtained through a stationary x-ray source. The fast temporal resolution in SSRDCT can be explained from the following three points. First, it is well known that simultaneous data acquisition from multiple source-detector chains can boost temporal resolution. The SSRDCT has three source-detector chains, which are expected to boost temporal resolution in a fashion similar to the DSR [15]–[17]. Second, the three distributed x-ray sources in SSRDCT are stationary, and scanning x-ray beams are realized electronically rather mechanically. The stationary sources will deliver the same benefit as that shown in the EBCT [13]. Compared to mechanical rotation, electronic sweeping allows the source-point to be moved at a far greater speed. Lastly, by removing the vacuum-based x-ray sources and their power electronics away from the mechanical gantry, what are left on the gantry will be the semiconductor-based detectors and their low-power electronics. The engineering challenge to spin a heavily loaded gantry will be gone. This design will not only reduce the mechanical complexity for the gantry but also could lead to a much faster gantry rotation speed (i.e. smaller T_{rot}) and hence an even higher temporal resolution in SSRDCT.

Historically, the high temporal resolution in the DSR and EBCT was not achieved without some drawbacks. Because the DSR had to employ a huge gantry to mount multiple source-detector chains, it came with large size and high cost, and only one such system has been ever built. The EBCT has an anode ring that encloses the patient body. To steer an electron beam to strike on the large target ring, the EBCT system also came with big size and high cost. The proposed SSRDCT is based on a conventional CT gantry, due to the stationary distributed x-ray source and the smaller detectors. The resulted system will be compact and will have a size similar to the current CT systems. As a result, the SSRDCT can achieve a high temporal resolution without the aforementioned drawbacks of DSR and EBCT. Additionally, the EBCT is also known for its other drawbacks from a stationary full-ring detector. A stationary full-ring detector, in addition to its high cost, also requires an offset in the z-direction between the source and detector planes to cover a typical scan angle of 220°. This configuration does not allow a typical anti-scatter grid collimator on the detector. In contrast, because the x-ray beams and the detectors are co-planar in the SSRDCT, detector anti-scatter grids can be still used. This is made possible because the rotating detectors are always at the opposite sides of the sweeping source-points. The sweeping source points can be realized by combining the electronic switching

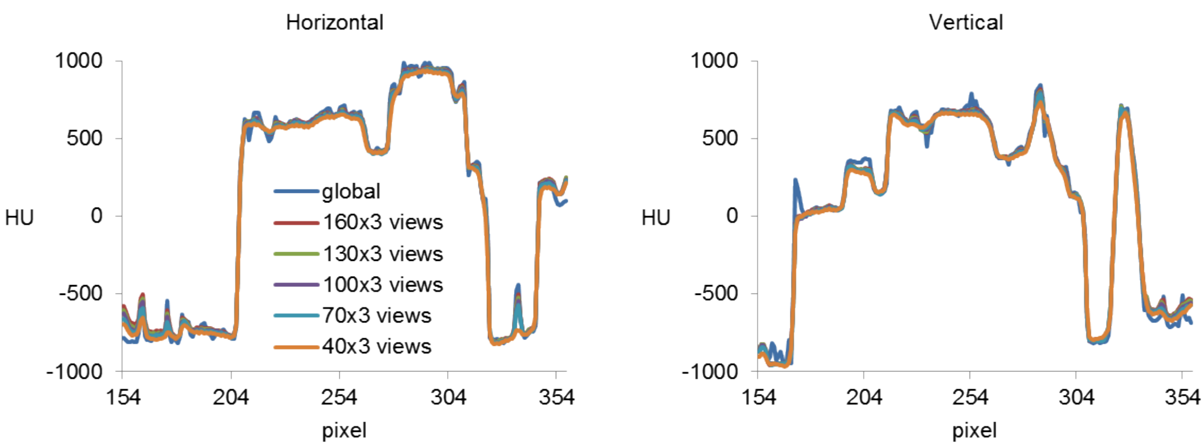

FIGURE 6. Two representative profiles along horizontal and vertical central lines for the images in Figure 5.

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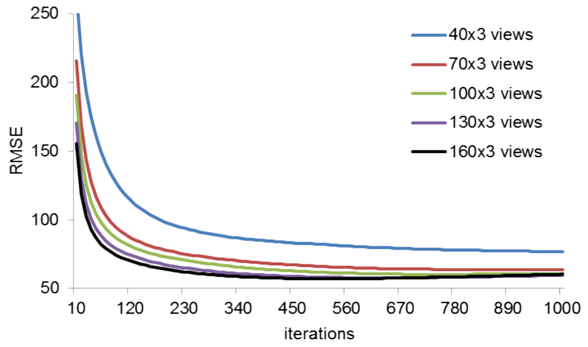

FIGURE 7. RMSEs vs. iteration numbers for different numbers of views at the half-scan mode.

of the gate voltages on the multiple cathodes and the local steering (i.e. steering electronically and/or magnetically the electron stream from a cathode to strike onto short focal track on the fixed anode during the exposure time for a projection).

The dose reduction arises from the interior scan and the CS-based interior reconstruction algorithm. The smaller an ROI, the narrower the beam, and thus less radiation. Narrower x-ray beams also reduce photon scattering to improve contrast resolution. The fewer views a CT scan acquires, the smaller number of x-ray beam assemblies will be required in a distributed x-ray source, and hence lower construction cost for the source. The distributed x-ray sources allow swift pulsing of the x-ray beams, which can eliminate the unnecessary x-ray exposures to save radiation dose. Therefore, the state-of-the-art few-view based interior tomography is highly synergistic with the distributed x-ray sources technology. To realize the benefit in lower radiation dose, we need to collimate the scanning x-ray beams from the stationary distributed x-ray sources during a CT scan. This can be realized by mounting three x-ray beam collimators on the rotating gantry and opposite to the three detectors. This arrangement can ensure that an x-ray beam emitted from the active source will be always well aligned with the corresponding collimator and detector.

The proposed SSRDCT architecture is not perfect and it has the following limitations. First, it has a small FOV (200mm), which is not sufficient to cover a full width of human body. Second, the critical components - the distributed x-ray sources – pose some engineering challenges. For the first limitation, the hospitals may need a dedicated heartcentered CT architecture because of the prevalence and morbidity of CVDs. The potential in higher temporal resolution and lower radiation dose is especially attractive for cardiac CT perfusion imaging studies. A heart-dedicated CT scanner, even though it can only cover 200mm FOV, can also find it suitable to evaluate other vital organs such as kidneys and livers. For the second limitation, while such sources are not available yet, recent progresses in CNT-based distributed x-ray sources provided a high possibility for their implementations. A linear x-ray tube with 75 CNT-based x-ray beams has been constructed to test some other interesting CT configurations [36]. Another linear x-ray tube with 31 individually controllable x-ray sources has been developed for stationary digital breast tomosynthesis [29]. It is our hope that the potential improvements in speed and dose of this SSRDCT architecture will promote the realization of such x-ray sources sooner.

Finally, other variants (in addition to the specific type in Figure 1) can be designed in the general frame of the SSRDCT architecture. An obvious extension is to use a stationary ring source that covers a full circle. This will make it possible for a full-scan. However, it will have higher cost on the source, require each detector to rotate at least 120°, and hence lower the temporal resolution. Another possibility is to use linear source arrays to replace curved source arrays. This design will lead to geometrical nonuniformity from view to view, but it could lower the cost for the distributed x-ray sources. Designing a new CT architecture is a complex task. At this stage, we mainly focused on identifying some feasible sampling geometry (coverage angles and view numbers etc.) and the corresponding image reconstruction algorithms. Some practical issues could have been not adequately investigated. For example, there may not be enough gap distance between the sources and detectors.

In conclusion, we present and evaluate a new SSRDCT architecture that is expected to provide higher temporal resolution and lower radiation dose. The SSRDCT architecture synergistically combines the few-view interior reconstruction algorithms and the distributed x-ray source technology. Although there may be some engineering challenges for its implementation, based on our analysis and simulation, the proposed SSRDCT architecture has a high potential to outperform the current CT architecture in both temporal resolution and radiation dose. According to the World Health Organization (WHO), the number of deaths from CVDs is estimated to be about 20 million in 2015 [37]. The heart-dedicated CT scanners based on the proposed SSRDCT architecture could be useful for our global mission to reduce deaths from CVDs.

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